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MRI RF-Induced Heating in Heterogeneous Human Body with Implantable Medical Device

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Abstract

Magnetic resonance imaging (MRI) radio frequency (RF)-induced heating is one of the most important concerns of MRI safety for patients, especially with orthopaedic healthcare products. In this chapter, numerical modelling and simulations were conducted to study the RF-induced heating within a 1.5T and 3T magnetic resonance (MR) environment. Numerical simulations were firstly conducted to study the difference between the cases of implantable medical devices inside the phantom and the human body. Then, numerical modelling were conducted to describe the difference of electromagnetic behaviours between the homogeneous phantom and heterogeneous human tissues. The MRI RF-induced heating due to an implantable medical device behaves significantly different in homogeneous media and in heterogeneous human body. For typical orthopaedic medical devices, such as healthcare products applied to shoulder, humerus, hip, femur and tibia, the properties of the RF-induced heating are different in general phantom and in human body. The hot spot location, as well as worst case configuration were evaluated and it was found that they were determined by the incident field and electromagnetic properties of medium. With further scaling, the RF-induced heating in human body for the orthopedic devices can be assessed by phantom studies.

Keywords: MRI, RF-induced heating, orthopedic implant, phantom

1. Introduction

Many of the MR-related injuries and the few fatalities that have occurred were the apparent result of failure to follow safety guidelines or of the use of inappropriate information related to the safety aspects of biomedical implants and devices [1–7]. The preservation of a safe MR environment requires constant attention to the care of patients and individuals with

metallic implants and devices, because the variety and complexity of these objects constantly changes [5–7]. Therefore, to guard against accidents in the MR environment, it is important to understand the risk associated with implantable medical devices which may cause potential problems.

The radiofrequency coils could send energy, in the form of electromagnetic radiation, into the human body. Since the energy is in the radio frequency range, the radiation is not ionizing. But it still can influence biological tissue. During MR procedures, the majority of the RF power transmitted for imaging or spectroscopy (especially for carbon decoupling) is transformed into heat within the patient's tissue as a result of resistive losses, through convection, conduction, radiation or evaporation [8–18]. Thus, a potential concern in MRI is the heating of the body during image acquisition.

To evaluate the RF-induced heating, the specific absorption rate (SAR) is applied to determine how much electromagnetic energy is absorbed by the body. SAR is typically expressed in unites of watts per kilogram, or W/kg. So the SAR could be defined as:

$$SAR(r) = \frac{\sigma}{2\rho} E^2(r) \quad (1)$$

where E is the total electric field and σ and ρ are the conductivity and density of biological tissue, respectively. The temperature rise in human body or phantom could be calculated by the total SAR according to the bio heat equation. SAR depends on the pulse sequence and the size, geometry, and conductivity of the absorbing object. To ensure participant safety, SAR in MRI studies is limited to minimize temperature increases.

The first study of human thermal responses to RF radiation-induced heating during an MR procedure was conducted by Schaefer et al. [19]. Temperature changes and other physiological parameters were assessed in volunteer subjects exposed to relatively high, whole-body averaged SARs (approximately 4.0 W/kg). The data indicated that there were no excessive temperature elevations or other deleterious physiological consequences related to the exposure to RF radiation [19].

However, for patients with medical implants, MRI-related RF induced heating is potentially problematic. The evaluation of heating for an implant or device is particularly challenging because of the many factors that affect temperature increase in these items. Variables that impact heating include the following: the specific type of implant or device; the electrical characteristic of the implant or device; the RF wavelength of the MR system; the type of transmit RF coil that is used (i.e., transmit head versus transmit body RF coil); the amount of RF energy delivered (i.e., the specific absorption rate, SAR); the landmark position or body part undergoing MRI relative to the transmit RF coil; and the orientation or configuration of the implant or device relative to the source of transmit RF coil.

In this chapter, it shows the importance of evaluation the MRI-related RF induced heating issues for patient with implantable medical devices. Generally, the estimation and measurement is based on *in-vitro* numerical simulation and experiment. And assessment methods could be separated into active and passive medical implants, respectively, due to the

configuration difference of these devices. With the help of the *in-vitro* evaluation methods, it provides a highly possible way to estimate the temperature increase for patient with implants or devices during MRI examination.

2. *In-vivo* and *in-vitro*

MRI may be contraindicated for a given patient primarily due to its potential risks associated with a metallic implant or device. Although many investigations have been performed using laboratory animals to determine thermoregulatory reactions to tissue heating associated with exposure to RF radiation, these experiments do not directly apply to the conditions that occur during MR procedures, nor can they be extrapolated to provide useful information for various reasons [20, 21]. For example, the pattern of RF absorption or the coupling of radiation to biological tissues is primarily dependent on the organism's size, anatomical features, duration of exposure, sensitivity of the involved tissues (e.g., some tissues are more "thermal sensitive" than others), and a myriad of other variables [14, 21, 22]. Furthermore, there is no laboratory animal that sufficiently mimics or simulates the thermoregulatory responses of an organism with the dimensions and specific responses to that of a human subject. Therefore, experimental results obtained in laboratory animals cannot be simply "scaled" or extrapolated to predict thermoregulatory or other physiological changes in human subjects exposed to RF radiation-induced heating during MR procedures [14, 15, 22], and. In consideration of the above, *in-vitro* testing is performed to assess the various MRI issues for implants and devices in order to properly characterize the possible risks.

One of *in-vitro* methods is to use standard American Society for Testing and Materials (ASTM) phantom. ASTM F2182-11A depicts the guideline to measure the RF heating induced by implanted medical devices in a standard phantom filled with gelled-saline which mimic the muscles [23]. Studies have been conducted to evaluate the RF heating induced by orthopedic implants. Commonly a phantom or homogenous media is used to mimic the environments as the implants locate in human body in experiments and/or numerical simulations [24–32].

Although the RF-induced heating evaluating method using the phantom filled with gelled-saline is widely used, it is obvious that the RF environment of a human body and a phantom filled with gelled-saline are quite different. The power deposition due to an implant for a given incident RF field is a function of the physical properties of the implant and electrical properties of the surrounding medium. Compared with homogeneous gelled-saline in phantom, human body is an inhomogeneous circumstance which includes different tissues with various permittivity and conductivity in a wide range. Hence, it is necessary to study a feasible guide with *in-vitro* phantom to assess the RF-induced heating in heterogeneous human body.

2.1. Human body: heterogeneous medium

With the development of computational electromagnetics, anatomical computer models of the human body have been used for nearly four decades for dosimetric applications in electromagnetics (EM) [33] and in medical radiation physics [34]. The most prominent numerical

| Name | Age (years) | Gender | Height (m) | Mass (kg) | BMI (kg/m²) |
|------|-------------|--------|------------|-----------|-------------|
| Duke | 34 | Male | 1.74 | 70 | 23.1 |

Table 1. The characteristics of the anatomical Duke model.

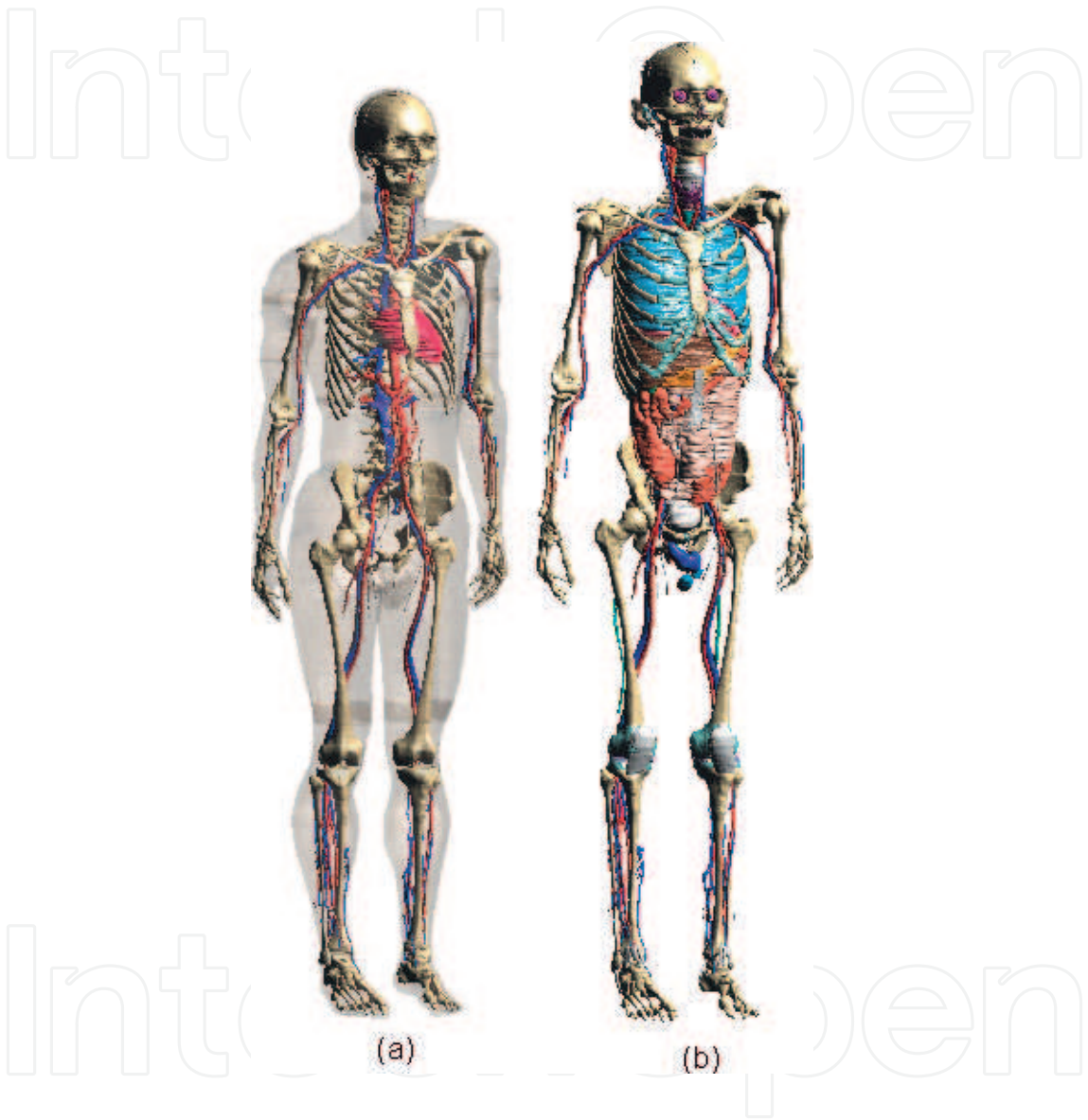


Figure 1. The segmented tissues and organs of anatomic body.

methods used in computational dosimetry of electromagnetic fields are based on finite-difference formulations of the underlying differential equations. For the simulation of both RF fields and induced tissue heating, the finite-difference time-domain (FDTD) method in its formulations by Yee [35] and Patankar [36] is applied to rectilinear grids to optimally handle large voxel models. The reconstructed human model used in this Chapter is from the Virtual Family [37]. It is based on high resolution magnetic resonance images of healthy volunteers. Seventy seven different tissue types were distinguished during the segmentation. Currently,

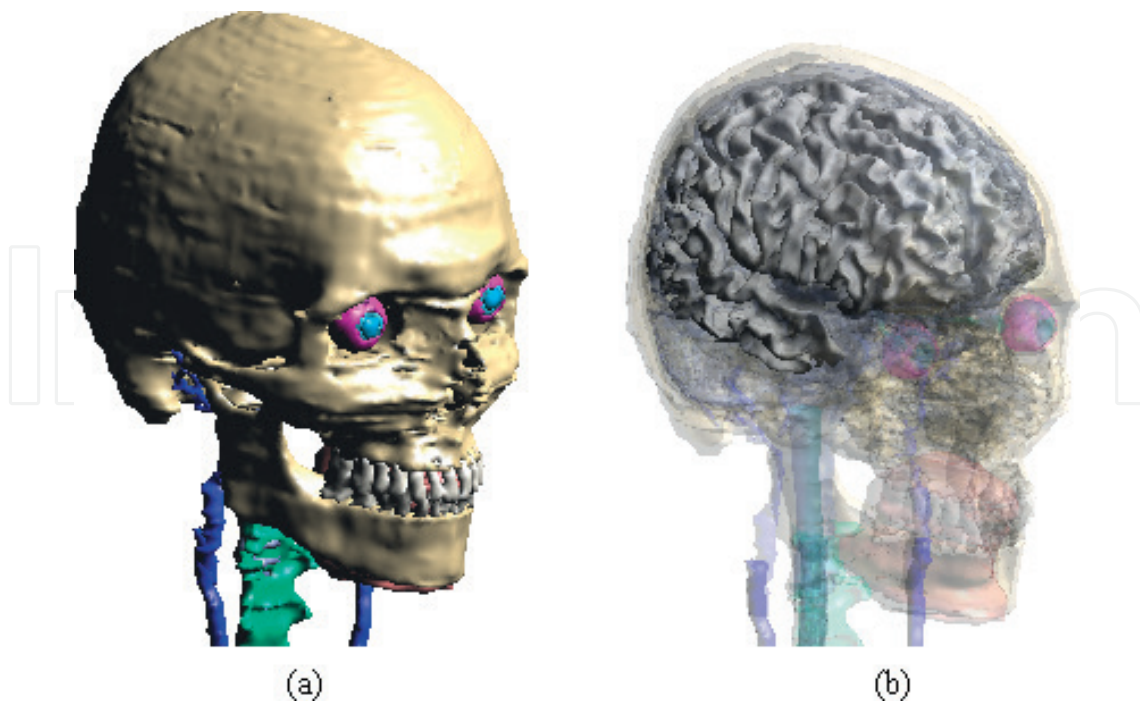


Figure 2. The segmented tissues and organs of anatomic brain.

| Tissue or organ | Electric conductivity (S/m) | | Relative permittivity | | Density (kg/m ³) |
|----------------------|-----------------------------|-------------|-----------------------|-------------|------------------------------|
| | 1.5 T/64 MHz | 3 T/128 MHz | 1.5 T/64 MHz | 3 T/128 MHz | |
| Adrenal gland | 0.778305 | 0.804166 | 73.9472 | 66.7839 | 1027.5 |
| Air internal | 0 | 0 | 1 | 1 | 1.2 |
| Artery | 1.20667 | 1.24863 | 86.4441 | 73.159 | 1049.75 |
| Bladder | 0.287352 | 0.298014 | 24.5943 | 21.8607 | 1035 |
| Blood vessel | 1.20667 | 1.24863 | 86.4441 | 73.159 | 1049.75 |
| Bone | 0.0595255 | 0.0673524 | 16.6812 | 14.7171 | 1908 |
| Brain gray matter | 0.510868 | 0.58673 | 97.4294 | 73.5204 | 1044.5 |
| Brain white matter | 0.291504 | 0.342151 | 67.8358 | 52.5338 | 1041 |
| Bronchi | 0.528415 | 0.559346 | 58.8896 | 50.5714 | 1101.5 |
| Bronchi lumen | 0 | 0 | 1 | 1 | 1.2 |
| Cartilage | 0.452103 | 0.488375 | 62.9145 | 52.9242 | 1099.5 |
| Cerebellum | 0.719003 | 0.829397 | 116.35 | 79.7377 | 1045 |
| Cerebrospinal fluid | 2.06597 | 2.14301 | 97.3124 | 84.0406 | 1007 |
| Commissure anterior | 0.291504 | 0.342151 | 67.8358 | 52.5338 | 1041 |
| Commissure posterior | 0.291504 | 0.342151 | 67.8358 | 52.5338 | 1041 |
| Connective tissue | 0.474331 | 0.498727 | 59.4892 | 51.8568 | 1525 |
| Cornea | 1.00058 | 1.05874 | 87.3779 | 71.4566 | 1050.5 |

| Tissue or organ | Electric conductivity (S/m) | | Relative permittivity | | Density (kg/m ³) |
|-----------------------|-----------------------------|-------------|-----------------------|-------------|------------------------------|
| | 1.5 T/64 MHz | 3 T/128 MHz | 1.5 T/64 MHz | 3 T/128 MHz | |
| Diaphragm | 0.688213 | 0.719235 | 72.2347 | 63.4948 | 1090.4 |
| Ear cartilage | 0.452103 | 0.488375 | 62.9145 | 52.9242 | 1099.5 |
| Ear skin | 0.43575 | 0.522704 | 92.1679 | 65.437 | 1109 |
| Epididymis | 0.884871 | 0.926404 | 84.5272 | 72.1279 | 1082 |
| Esophagus | 0.877842 | 0.912807 | 85.8204 | 74.895 | 1040 |
| Esophagus lumen | 0 | 0 | 1 | 1 | 1.2 |
| Eye lens | 0.28588 | 0.312684 | 50.3392 | 42.7911 | 1075.5 |
| Eye sclera | 0.882673 | 0.917665 | 75.2998 | 64.9991 | 1032 |
| Eye vitreous humor | 1.50315 | 1.50536 | 69.1264 | 69.0619 | 1004.5 |
| Fat | 0.0661558 | 0.0697299 | 13.6436 | 12.3711 | 911 |
| Gall bladder | 1.48179 | 1.5764 | 105.443 | 88.8995 | 928 |
| Heart lumen | 1.20667 | 1.24863 | 86.4441 | 73.159 | 1049.75 |
| Heart muscle | 0.678423 | 0.766108 | 106.514 | 84.2573 | 1080.8 |
| Hippocampus | 0.510868 | 0.58673 | 97.4294 | 73.5204 | 1044.5 |
| Hypophysis | 0.778305 | 0.804166 | 73.9472 | 66.7839 | 1053 |
| Hypothalamus | 0.778305 | 0.804166 | 73.9472 | 66.7839 | 1053 |
| Intervertebral disc | 0.452103 | 0.488375 | 62.9145 | 52.9242 | 1099.5 |
| Kidney cortex | 0.741316 | 0.852313 | 118.556 | 89.6168 | 1049 |
| Kidney medulla | 0.741316 | 0.852313 | 118.556 | 89.6168 | 1044 |
| Large intestine | 0.638152 | 0.705214 | 94.6639 | 76.5722 | 1088 |
| Large intestine lumen | 0.688213 | 0.719235 | 72.2347 | 63.4948 | 1045.2 |
| Larynx | 0.452103 | 0.488375 | 62.9145 | 52.9242 | 1099.5 |
| Liver | 0.447984 | 0.510897 | 80.5595 | 64.2507 | 1078.75 |
| Lung | 0.288977 | 0.315616 | 37.1022 | 29.4677 | 394 |
| Mandible | 0.0595255 | 0.0673524 | 16.6812 | 14.7171 | 1908 |
| Marrow | 0.154335 | 0.162021 | 16.4355 | 13.5377 | 1028.5 |
| Medulla oblongata | 0.719003 | 0.829397 | 116.35 | 79.7377 | 1045.5 |
| Meniscus | 0.452103 | 0.488375 | 62.9145 | 52.9242 | 1099.5 |
| Midbrain | 0.719003 | 0.829397 | 116.35 | 79.7377 | 1045.5 |
| Mucosa | 0.488039 | 0.544202 | 76.7233 | 61.5852 | 1102 |
| Muscle | 0.688213 | 0.719235 | 72.2347 | 63.4948 | 1090.4 |
| Nerve | 0.312174 | 0.353802 | 55.0621 | 44.0653 | 1075 |
| Pancreas | 0.778305 | 0.804166 | 73.9472 | 66.7839 | 1086.5 |
| Patella | 0.0595255 | 0.0673524 | 16.6812 | 14.7171 | 1908 |

| Tissue or organ | Electric conductivity (S/m) | | Relative permittivity | | Density (kg/m ³) |
|-----------------------|-----------------------------|-------------|-----------------------|-------------|------------------------------|
| | 1.5 T/64 MHz | 3 T/128 MHz | 1.5 T/64 MHz | 3 T/128 MHz | |
| Penis | 0.429311 | 0.478934 | 68.6368 | 55.9888 | 1101.5 |
| Pharynx | 0 | 0 | 1 | 1 | 1.2 |
| Pineal body | 0.778305 | 0.804166 | 73.9472 | 66.7839 | 1053 |
| Pons | 0.719003 | 0.829397 | 116.35 | 79.7377 | 1045.5 |
| Prostate | 0.884871 | 0.926404 | 84.5272 | 72.1279 | 1045 |
| SAT | 0.0661558 | 0.0697299 | 13.6436 | 12.3711 | 911 |
| Skin | 0.43575 | 0.522704 | 92.1679 | 65.437 | 1109 |
| Skull | 0.0595255 | 0.0673524 | 16.6812 | 14.7171 | 1908 |
| Small intestine | 1.59145 | 1.69285 | 118.363 | 87.9725 | 1030 |
| Small intestine lumen | 0.688213 | 0.719235 | 72.2347 | 63.4948 | 1045.2 |
| Spinal cord | 0.312174 | 0.353802 | 55.0621 | 44.0653 | 1075 |
| Spleen | 0.743914 | 0.835186 | 110.559 | 82.8917 | 1089 |
| Stomach | 0.877842 | 0.912807 | 85.8204 | 74.895 | 1088 |
| Stomach lumen | 0.688213 | 0.719235 | 72.2347 | 63.4948 | 1045.2 |
| Teeth | 0.0595255 | 0.0673524 | 16.6812 | 14.7171 | 2180 |
| Tendon ligament | 0.474331 | 0.498727 | 59.4892 | 51.8568 | 1142 |
| Testis | 0.884871 | 0.926404 | 84.5272 | 72.1279 | 1082 |
| Thalamus | 0.510868 | 0.58673 | 97.4294 | 73.5204 | 1044.5 |
| Thymus | 0.778305 | 0.804166 | 73.9472 | 66.7839 | 1023 |
| Thyroid gland | 0.778305 | 0.804166 | 73.9472 | 66.7839 | 1050 |
| Tongue | 0.652145 | 0.687137 | 75.2998 | 64.9991 | 1090.4 |
| Trachea | 0.528415 | 0.559346 | 58.8896 | 50.5714 | 1080 |
| Trachea lumen | 0 | 0 | 1 | 1 | 1.2 |
| Ureter Urethra | 0.429311 | 0.478934 | 68.6368 | 55.9888 | 1101.5 |
| Vein | 1.20667 | 1.24863 | 86.4441 | 73.159 | 1049.75 |
| Vertebrae | 0.0595255 | 0.0673524 | 16.6812 | 14.7171 | 1908 |

Table 2. The electromagnetic properties of the segmented tissues and organs.

the models are being widely applied in several studies on electromagnetic exposure, device optimization and medical applications. **Table 1** shows the characteristics of the anatomical model. Duke model is an anatomical model of adult male which is shown in **Figures 1** and **2**. And **Table 2** shows the segmented tissues and organs of the model, as well as the electromagnetic properties.

2.2. ASTM phantom: *in-vitro* measurement

The standard F2182 describe a test method for measurement of RF induced heating on or near passive implants and its surrounding during MRI procedure. A design of phantom container is introduced in the standard with its dimension shown in **Figure 3**. The material of phantom container are electrical insulators and non-magnetic and non-metallic. The phantom container is filled with a gelled-saline which has a relative permittivity $\epsilon_r = 80.4$ and conductivity of $\sigma = 0.47$ S/m. In order to have a great conductivity and viscosity, a suitable gelled saline should be made with 1.32 g/L NaCl and 10 g/L polyacrylic acid (PAA) in water. Numerical simulations indicate that the maximum electric field inside the ASTM phantom is at mid-axial plane about 2 cm away from the vertical phantom side wall. To maximize the heating, and thereby maximizing the signal-to-noise ratio, we placed the implants at this location.

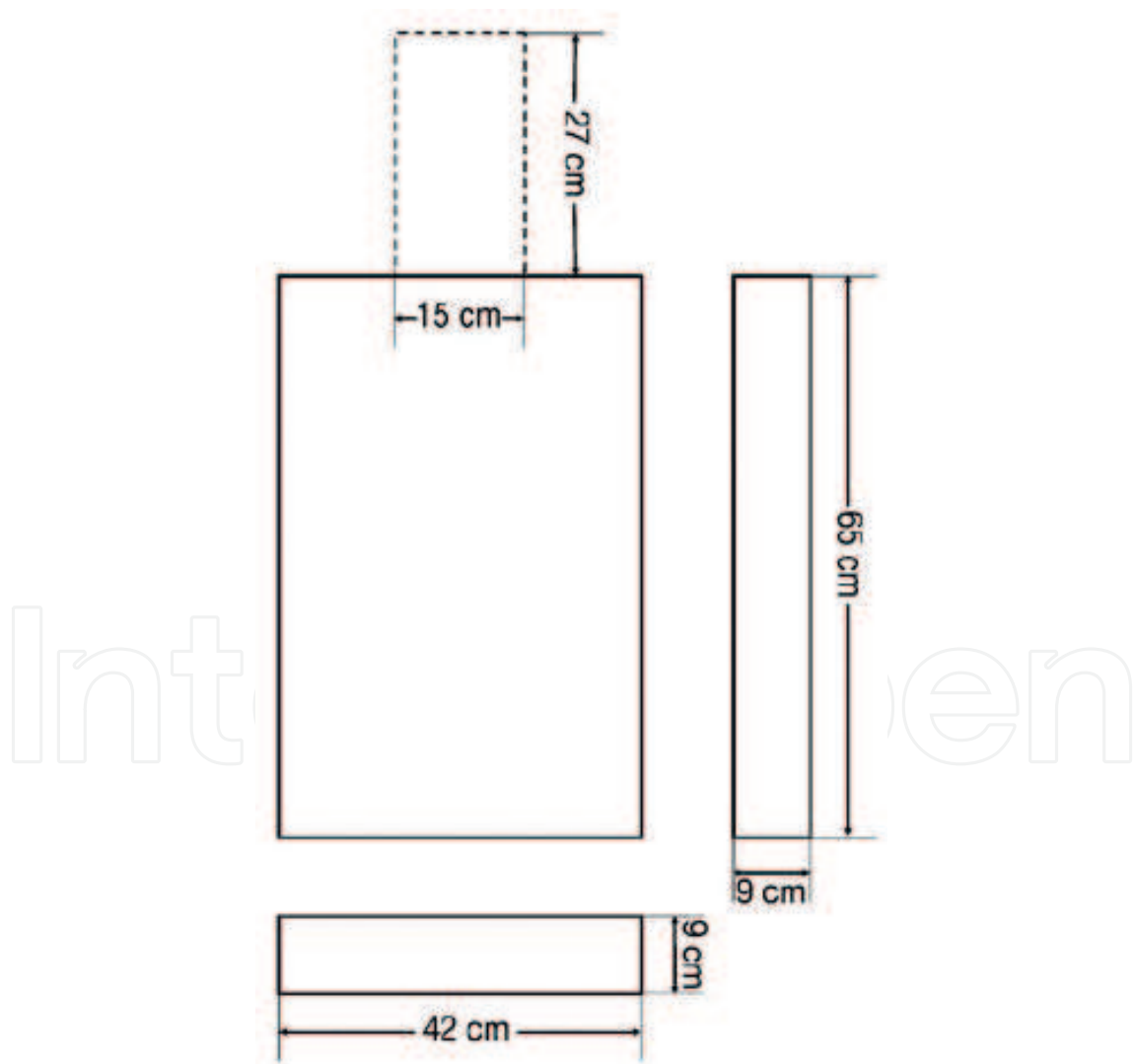


Figure 3. The structure and dimension of standard ASTM phantom.

A generic RF transmit body coil is developed and shown in **Figure 4**. The upper plots represent a 1.5 T RF coil and the lower two plots represent a 3 T RF coil. A physical coil is usually difficult to model and it takes much longer simulation time to reach the steady state of the simulation. It has been shown that using a non-physical coil could significantly reduce the simulation time while providing the same result as that from a physical coil. Thus, rather than modeling the exact physical coil, the non-physical coils were modeled in this study. The two coils have the same dimensions, and both have 8 rungs. The diameter of the RF coil is 63 cm, and the height of the RF coil is 65 cm. The eight parallel lines or the rungs are one dimensional line current excitation. The end rings on top and bottom of the RF coils are tuning capacitors which are also modelled as one dimensional line segments.

The capacitance value is determined from several broadband simulations so that the second highest resonant frequency was adjusted to 64 MHz for 1.5-T and 128 MHz for 3-T systems. The detailed steps are: set an initial capacitance value for all capacitors on end rings and add a broadband pulse signal on one single rung. The other seven rungs are modeled as zero ohm resistors. After the simulation is finished, the power spectrum is extracted. If the second highest resonant frequency is not at appropriate resonant frequencies, the capacitance needs to be adjusted. From this study, the capacitance for the end ring tuning capacitor values is 7.2 pF for 1.5-T RF coil and 1.3 pF for 3-T RF coil.

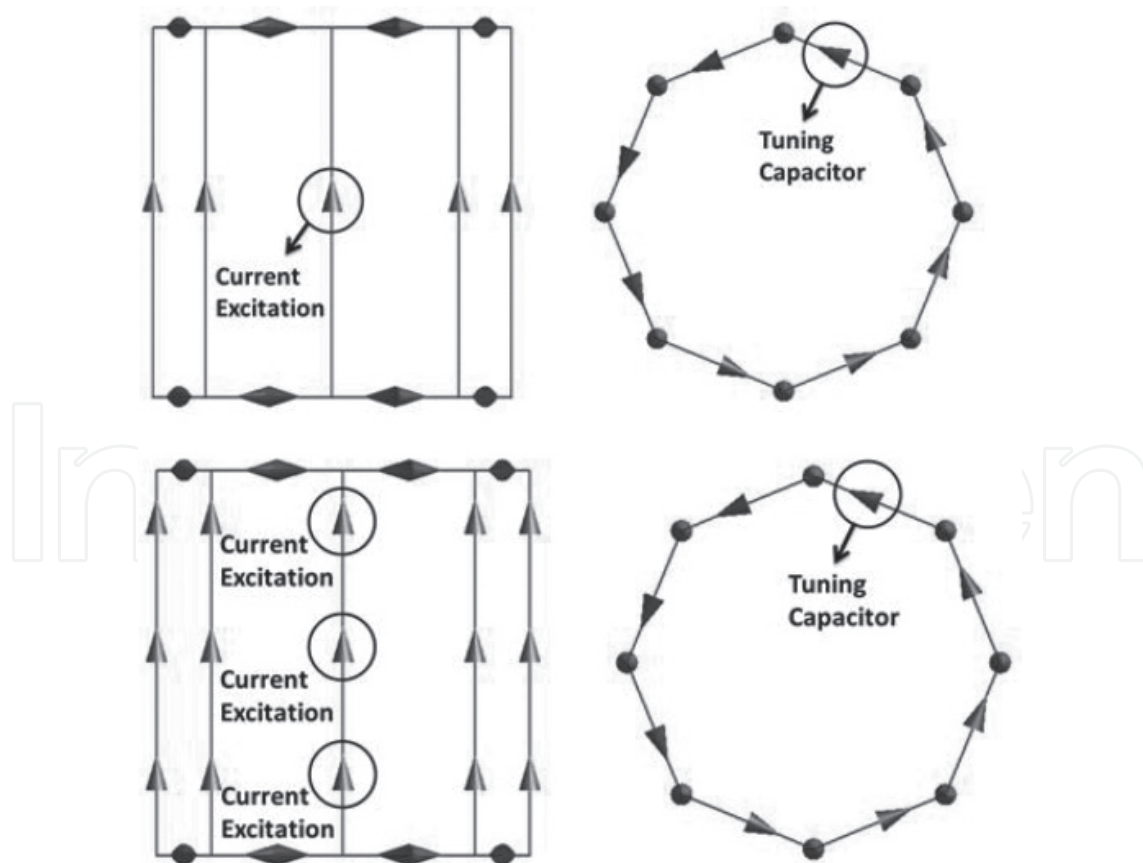


Figure 4. The generic coil model of 1.5-T RF coil (top) and 3-T RF coil (bottom) in SEMCAD X.

3. Passive implantable medical device

Any device intended to be totally or partially introduced into the human body through surgical intervention and intended to remain in place after the procedure for at a long-term duration is considered as an implantable device. Passive devices in terms of their form of operation can be classified as device used for transportation and storage of pharmaceutical liquid, device for alteration of blood, body fluids, medical dressing, surgical instruments; reusable surgical instruments, disposable aseptic device, implantable device, device for contraception and birth control, device for sterilization and cleaning, patient care device, in vitro diagnostic reagent, as well as other passive contacting device or passive supplementary device.

In this chapter, three typical categories of orthopedic implantable devices, bone plate system, hip prostheses and tibia intramedullary nails, are selected for MRI related RF induced heating study which are shown in **Figures 5–7**. The configuration of each implantable device is shown

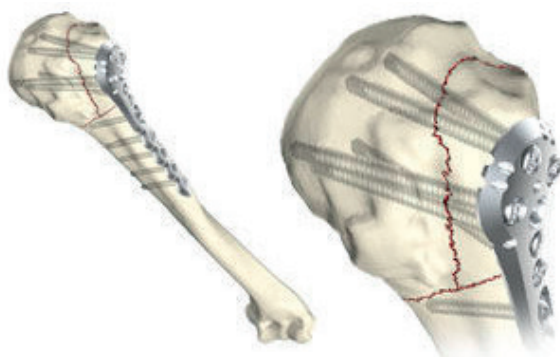


Figure 5. The bone plant system of AxSOS system from Stryker®.



Figure 6. The hip prostheses of Excia® T from Aesculap®.



Figure 7. The tibia intramedullary nails of PROtect™ from Depuy Synthes.

in figure. For bone fragment compression plate, it is designed to offer multiple compression and reconstruction plating options for the treatment of bone fractures. The application of hip prostheses is related to hip revision and arthroplasty. As for intramedullary nails, they are characterized by the anatomic shape, which is intended to replicate the natural anatomic shape of the bones. They have been designed to help restore the shape of the bone and treat the fractured bones.

4. Numerically evaluate RF-induced heating

4.1. FDTD method

In this numerical investigation, we use the finite difference time domain (FDTD) based SEMCAD X 14.8 (SPEAG) simulation platform. Graphics processing unit (GPU) hardware acceleration was achieved using the SPEAG CUDA library with Tesla C2075 graphic card which is can handle up to 90 million cells. To assure convergence of the numerical simulations, the simulation time was set to 20 periods for each simulation. Additionally, the convergence was checked after the simulations were finished. The material of orthopedic devices is set to perfect electric conductor (PEC), and all the numerical results are normalized to a whole body average SAR of 2 W/kg. The SAR distribution is studied for each case.

4.2. Bone plate system

To ensure a comprehensive comparison, the 1g local average peak SAR value at device is extracted for each configuration of femur and humerus system. **Tables 3** and **4** show the value for femur and humerus system. For each numerical result, whole-body average SAR is normalized to 2 W/kg. Since the interaction between RF induced field and implant is dependent on the physical structure of device, the heating effect variations related to the length of plate and screw are studied separately. For femur system, the plate length varies from 100 to 300 mm, and the screw length changes from 10 to 32 mm. For humerus system, the screw dimension is the same as femur system. But the plate length varies only from 100 to 250 mm due to the limit of bone structure. The plate length is studied at first for minimum and maximum screw length. Then the screw length is investigated under the worst case of plate length study which has the highest 1g average peak SAR value for *in-vivo* simulation. **Figures 8–13** show the results which are corresponding to femur and humerus plate system. The solid and dash curve and indicate the *in-vivo* and *in-vitro* results, respectively.

| Plate length (mm) | | Screw length (mm) | 1.5 T/64 MHz | | 3 T/128 MHz | |
|-------------------|----------|-------------------|---|--|---|--|
| | | | <i>In-vivo</i> SAR _{1g} (W/kg) | <i>In-vitro</i> SAR _{1g} (W/kg) | <i>In-vivo</i> SAR _{1g} (W/kg) | <i>In-vitro</i> SAR _{1g} (W/kg) |
| 100 | | 10 | 64.20 | 125.97 | 79.75 | 80.90 |
| 150 | | 10 | 94.82 | 178.62 | 74.52 | 64.01 |
| 175 | | 10 | 107.00 | 190.44 | 68.50 | 50.20 |
| 200 | | 10 | 116.65 | 188.87 | 63.37 | 44.69 |
| 225 | | 10 | 117.00 | 185.04 | 63.10 | 37.72 |
| 250 | | 10 | 123.00 | 169.75 | 61.23 | 37.27 |
| 275 | | 10 | 117.00 | 149.81 | 53.80 | 35.33 |
| 300 | | 10 | 105.00 | 134.91 | 42.37 | 38.53 |
| 100 | | 32 | 85.02 | 100.22 | 88.26 | 71.90 |
| 150 | | 32 | 108.17 | 135.90 | 55.48 | 47.74 |
| 200 | | 32 | 104.91 | 140.27 | 51.56 | 41.59 |
| 250 | | 32 | 111.73 | 123.17 | 48.94 | 37.09 |
| 300 | | 32 | 79.94 | 128.93 | 27.77 | 40.70 |
| 250(1.5 T) | 100(3 T) | 15 | 120.00 | 150.91 | 79.30 | 72.19 |
| 250(1.5 T) | 100(3 T) | 20 | 121.00 | 135.71 | 76.20 | 68.00 |
| 250(1.5 T) | 100(3 T) | 25 | 123.00 | 132.15 | 64.2 | 66.52 |

Table 3. Peak 1g averaged SAR of femur system for *in-vivo* and *in-vitro* cases.

| Plate length (mm) | Screw length (mm) | 1.5 T/64 MHz | | 3 T/128 MHz | |
|-------------------|-------------------|---|--|---|--|
| | | <i>In-vivo</i> SAR _{1g} (W/kg) | <i>In-vitro</i> SAR _{1g} (W/kg) | <i>In-vivo</i> SAR _{1g} (W/kg) | <i>In-vitro</i> SAR _{1g} (W/kg) |
| 100 | 10 | 38.02 | 135.61 | 65.47 | 90.51 |
| 150 | 10 | 63.74 | 192.14 | 94.68 | 59.90 |
| 200 | 10 | 69.47 | 204.57 | 81.84 | 45.54 |
| 250 | 10 | 104.57 | 193.61 | 86.73 | 41.50 |
| 100 | 32 | 30.70 | 92.17 | 63.46 | 64.6 |
| 150 | 32 | 54.10 | 124.76 | 85.39 | 53.70 |
| 200 | 32 | 55.90 | 161.97 | 75.19 | 43.39 |
| 250 | 32 | 109.00 | 156.38 | 108.92 | 42.51 |
| 250 | 15 | 95.53 | 161.64 | 77.82 | 35.33 |
| 250 | 20 | 91.02 | 163.55 | 72.91 | 38.94 |
| 250 | 25 | 88.27 | 164.54 | 68.40 | 41.32 |

Table 4. The peak 1g average SAR value of humerus system for *in-vivo* and *in-vitro* cases.

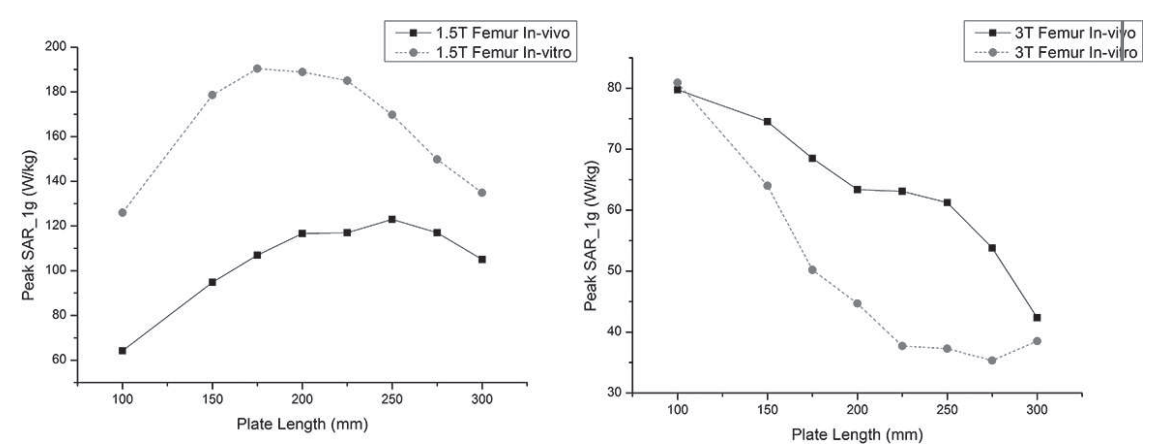


Figure 8. The femur bone plate length study of 10 mm screw for 1.5 T (left) and 3 T (right).

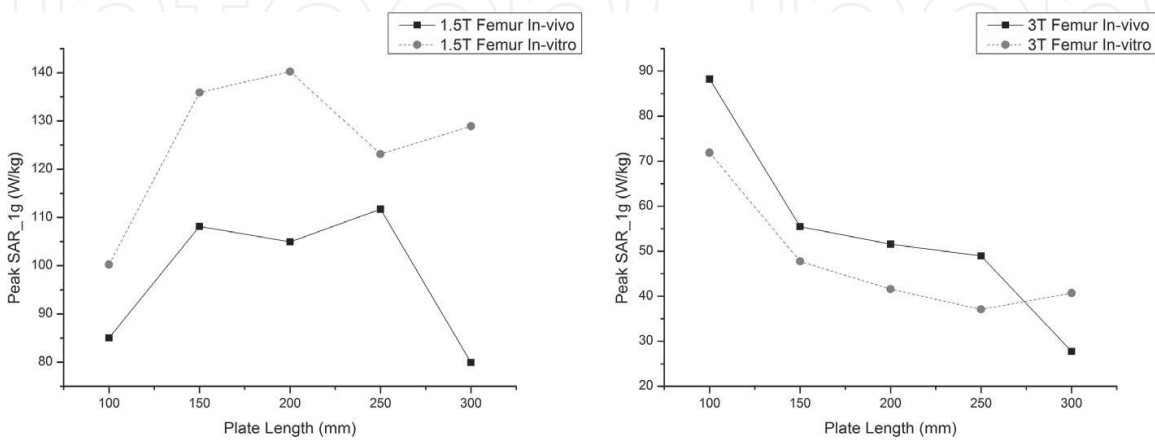


Figure 9. The femur bone plate length study of 32 mm screw for 1.5 T (left) and 3 T (right).

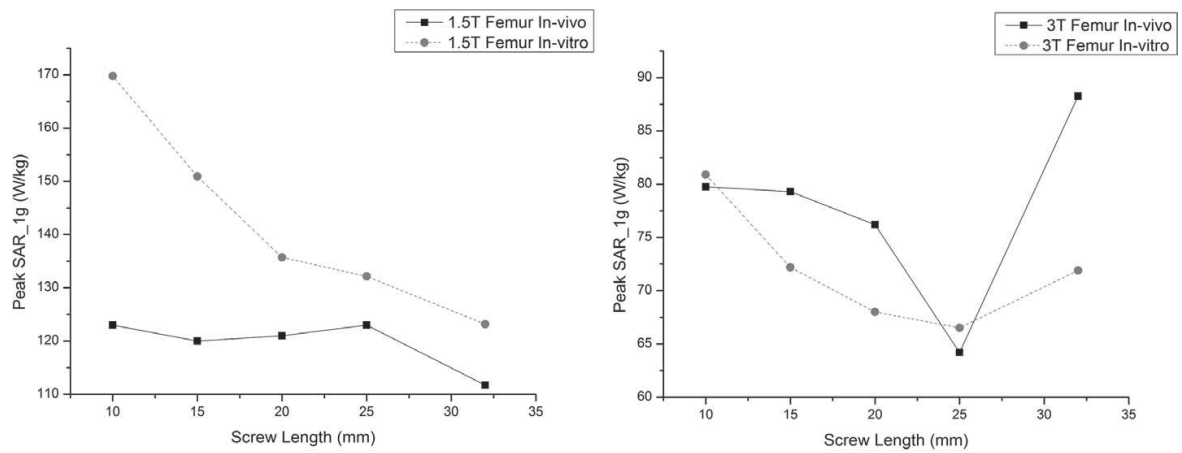


Figure 10. The femur screw length study for 1.5 T (left) and 3 T (right).

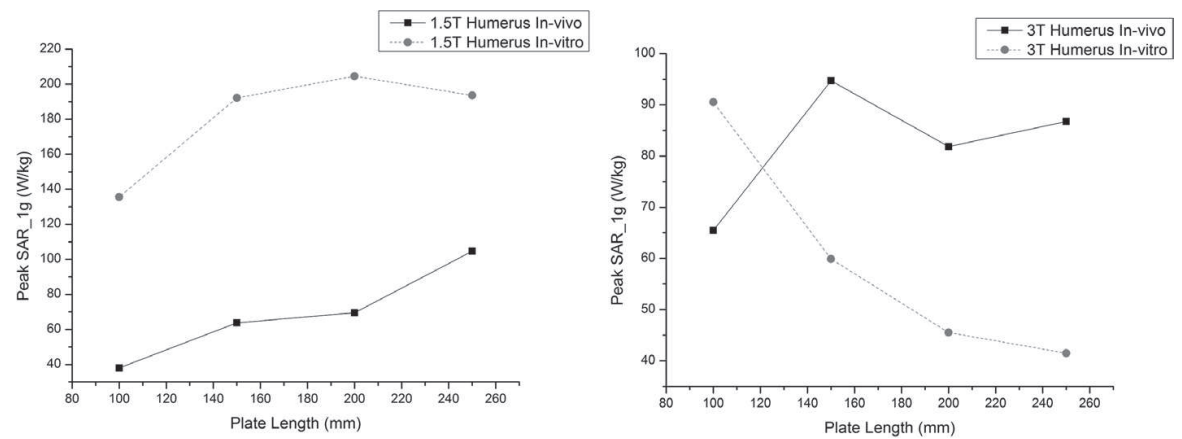


Figure 11. The humerus bone plate length study of 10 mm screw for 1.5 T (left) and 3 T (right).

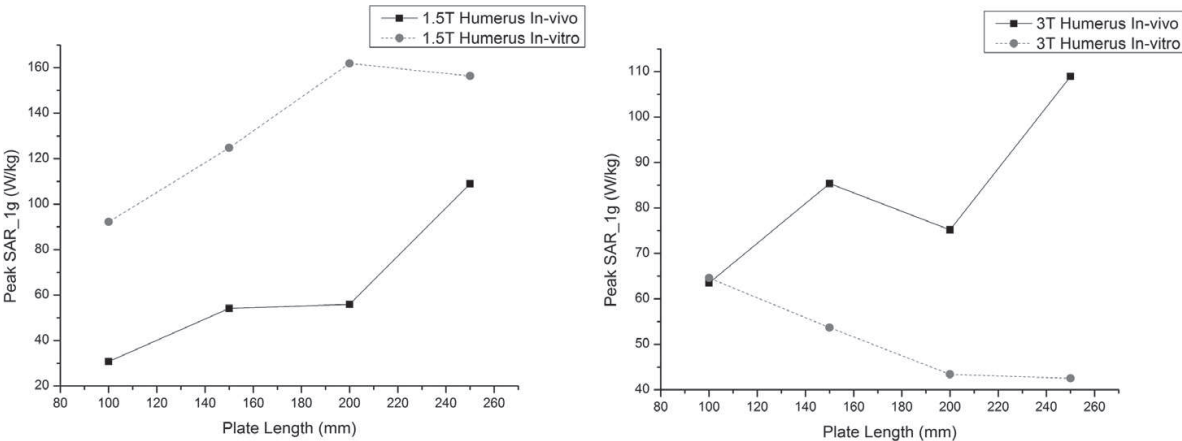


Figure 12. The humerus bone plate length study of 32 mm screw for 1.5 T (left) and 3 T (right).

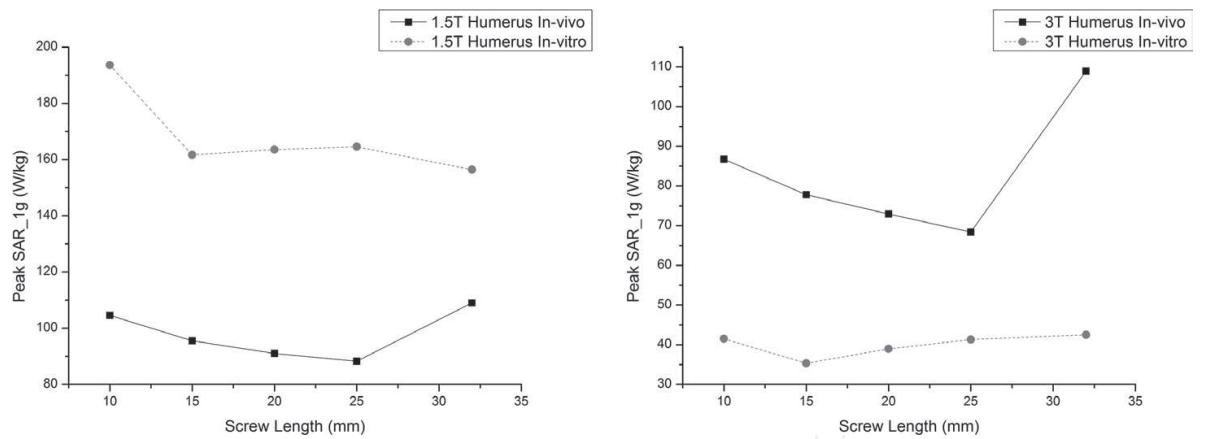


Figure 13. The humerus screw length study for 1.5 T (left) and 3 T (right).

4.3. Hip prostheses

For hip prostheses, the 1g average local peak SAR value at device is also extracted for each configuration. **Table 5** shows the value for hip system of various dimensions. The height of hip prostheses stem ranges from 100 to 170 mm. For *in-vivo* simulation, the stem is inserted into the bone marrow. And for the trochanter region, the hip prostheses is touching with soft tissue and muscle. **Figure 14** represents the results of hip prostheses. The solid and dash curve and indicate the *in-vivo* and *in-vitro* results, respectively.

4.4. Tibia intramedullary nails

The 1g average local peak SAR value at device is also extracted for each configuration of tibia intramedullary nails. The length of stem ranges from 255 to 360 mm. The entire nail

| Stem height (mm) | 1.5 T/64 MHz | | 3 T/128 MHz | |
|------------------|---|--|---|--|
| | <i>In-vivo</i> SAR _{1g} (W/kg) | <i>In-vitro</i> SAR _{1g} (W/kg) | <i>In-vivo</i> SAR _{1g} (W/kg) | <i>In-vitro</i> SAR _{1g} (W/kg) |
| 100 | 83.2764 | 270.843 | 43.4746 | 55.4143 |
| 110 | 81.5782 | 260.502 | 35.8247 | 56.3111 |
| 120 | 77.8568 | 248.116 | 24.4627 | 57.0498 |
| 130 | 74.9259 | 247.635 | 21.9093 | 60.352 |
| 140 | 64.5142 | 237.51 | 16.3059 | 61.7026 |
| 150 | 62.8772 | 205.328 | 13.1482 | 63.2055 |
| 160 | 59.6273 | 221.781 | 11.5599 | 63.9114 |
| 170 | 55.0469 | 213.602 | 11.1877 | 63.8915 |

Table 5. Peak 1g average SAR of hip prostheses system for *in-vivo* and *in-vitro* cases.

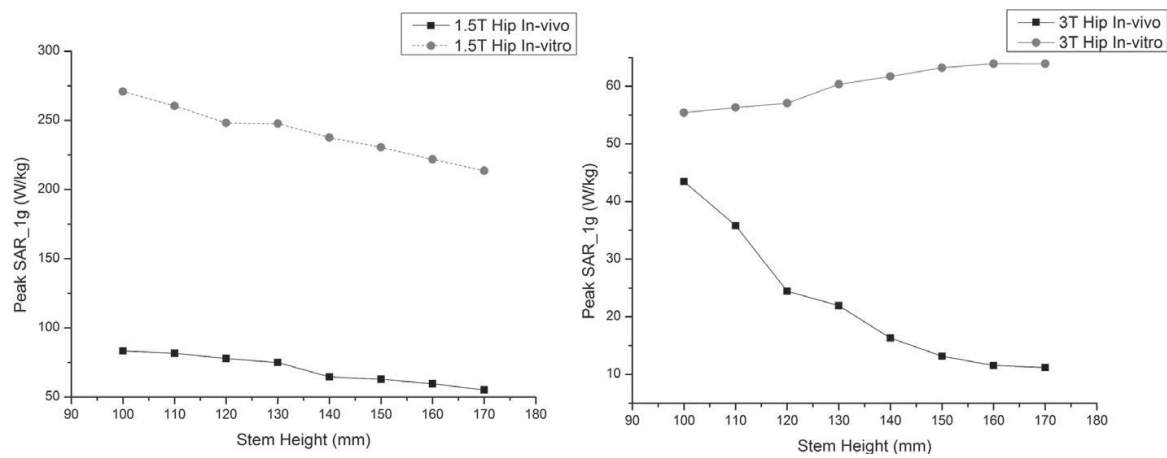


Figure 14. The hip prostheses stem length study for 1.5 T (left) and 3 T (right).

| Nail length (mm) | 1.5 T/64 MHz | | 3 T/128 MHz | |
|------------------|---|--|---|--|
| | <i>In-vivo</i> SAR _{1g} (W/kg) | <i>In-vitro</i> SAR _{1g} (W/kg) | <i>In-vivo</i> SAR _{1g} (W/kg) | <i>In-vitro</i> SAR _{1g} (W/kg) |
| 255 | 77.6968 | 136.331 | 93.5532 | 55.4068 |
| 270 | 77.6346 | 129.249 | 93.2341 | 53.6038 |
| 285 | 78.9892 | 122.28 | 89.1767 | 49.8946 |
| 300 | 82.6751 | 115.745 | 91.5648 | 46.8385 |
| 315 | 81.7577 | 109.28 | 92.7346 | 44.0529 |
| 330 | 74.7444 | 103.319 | 88.9989 | 41.6544 |
| 345 | 66.6095 | 97.3232 | 90.7795 | 39.3275 |
| 360 | 67.3469 | 91.8168 | 90.1809 | 37.1875 |

Table 6. Peak 1g average SAR of tibia nails system for *in-vivo* and *in-vitro* cases.

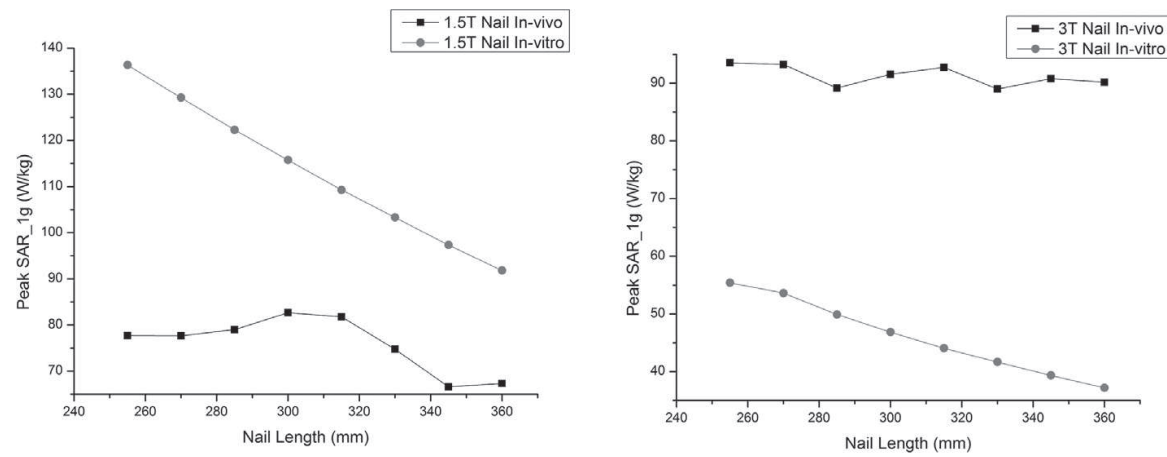


Figure 15. The nail length study for 1.5 T (left) and 3 T (right).

is penetrated into the bone marrow. The four screws are inserted perpendicularly through the nail and bone. **Table 6** shows the value for nail system of various dimensions. **Figure 15** represents the results of tibia intramedullary nails. The solid and dash curve and indicate the *in-vivo* and *in-vitro* results, respectively.

5. Summary

From the comparison between *in-vitro* and *in-vivo* simulations, the RF-induced heating are different because of the variance of incident electric field and surrounding medium. For incident field study, the antenna resonance effect would mainly lead to a heating issue for both *in-vitro* and *in-vivo* situation. Although the wavelength of human muscle and gelled-saline nearly equals to each other, due to the variance of incident RF field, the device dimension causing the resonance would be different. Hence, the trend of peak 1g average SAR value along with plate length is unlike from *in-vitro* to *in-vivo* circumstance. Additionally, when the screw is inserted across the human bone into the muscle, a huge amount of power would dissipated to the human tissue through the screw tip so that induce a large peak SAR value.

Based on the comparison result, conservatively, the *in-vitro* method, such as ASTM phantom, could be used to assess RF-induced heating. However, to accurately assess the RF-induced heating in heterogeneous human body with implantable medical device, due to the limit of homogeneous ASTM phantom, it still needs some improvement to handle several particular cases, especially, when the implantable device is penetrating through various human tissues and organs.

Disclaimer

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